[0051] To evaluate spatial and temporal gait symmetry, the differences in step lengths (L_{sym}) (m) and step times (T_{sym}) (s) from the affected to the unaffected side were compared for each of the three control conditions (n=20). The results are set forth in Table IV below.

TABLE IV

	L_{sym}(m)		T_{sym}(s)	
	Self-selected	Slow	Self-selected	Slow
Zero Impedance	0.08 ± 0.07	0.09 ± 0.09	0.09 ± 0.07	0.15 ± 0.16
Constant Impedance	0.04 ± 0.06	0.02 ± 0.08	0.07 ± 0.05	0.04 ± 0.12
Variable Impedance	0.02 ± 0.07	0.00 ± 0.07	0.02 ± 0.09	0.01 ± 0.16

[0052] Both L_{sym} and T_{sym} for the variable-impedance controller were significantly smaller than the zero impedance controller for both the self-selected and slow gait speeds ($p<0.05$). The zero and constant impedance conditions were significantly different for the slow gait speed ($p<0.05$). For the fast gait speed, a comparison was not possible because the step length for both sides could not be calculated for a single walking cycle.

[0053] An active ankle foot orthosis is provided in accordance with aspects of the present invention. Zero, constant, and variable-impedance control strategies were evaluated on two persons suffering from unilateral drop foot gait. It was found that actively adjusting joint impedance in response to walking phase and forward speed reduces the occurrence of slap foot, and provides for swing phase ankle kinematics more closely resembling normals as compared to the zero and constant impedance control schemes. Furthermore, it was found that a variable-impedance control allows for greater powered plantar flexion compared to a conventional constant stiffness approach where a dorsiflexion spring impedes powered plantar flexion movements during late stance.

[0054] Although the major complications of drop foot are reduced with a variable-impedance control, the findings do not support the hypothesis that changing orthotic joint impedance will result in a more symmetric gait between affected and unaffected legs in unilateral drop foot gait. To test the hypothesis, spatial and temporal gait symmetry was evaluated according to the difference in step lengths and times between affected and unaffected sides. When using the variable-impedance control, the difference in step time and step length was not significantly different from that measured with the constant impedance control condition. However, for both gait speeds analyzed, the variable-impedance controller did improve spatial and temporal gait symmetry compared to the zero impedance control condition, whereas the constant impedance control did not.

[0055] The CP stiffness was optimized within each gait speed range, or time of contact bin. After the variable-impedance controller adapted CP stiffness across gait speed, the final stiffness at the slow speed was 36% less, and at the fast speed, 57% greater than at the self-selected speed. Thus, from slow to fast speeds, stiffness increased more than two-fold. A constant stiffness spring tuned only to the self-selected speed allowed slap foot to occur at fast walking speeds (**FIG. 6**). It also made the ankle too stiff during slow walking, reducing the angular rotation of the ankle during controlled plantar flexion movements in early stance.

[0056] The primary concern for both the drop foot participants in the study was catching their toe during swing and losing their balance. With constant swing phase impedance, both users caught their toe at the fast gait speed. This was not surprising given the fact that, for normal gait, the amount of time to lift the foot and achieve toe clearance was found to decrease by a factor of two from slow to fast speeds. To achieve this time decrease with the AAFO **10**, a four-fold increase in swing joint stiffness was necessary (Table II). Thus, changing orthotic joint impedance with gait speed, in order to lift the toe during swing, appears to be a desired control feature of the variable-impedance AAFO **10**.

[0057] Normal ankle function has been modeled as a linear spring during controlled plantar flexion, and as a non-linear, stiffening spring during controlled dorsiflexion. Throughout the swing phase, the ankle has been represented by a linear torsional spring and damper. Given these differences in ankle function within a single gait cycle, an assistive ankle device, acting in parallel with the human ankle-foot complex, should ideally change its impedance in response to walking phase. To this end, a state controller was used in the AAFO **10**, and joint impedance was modulated in response to walking phase.

[0058] During the controlled plantar flexion phase of walking, or Contact **1**, a linear torsional spring control was employed where the stiffness was adjusted to prevent slap foot. From mid-stance to pre-swing, or the Contact **2** state, a zero impedance control was implemented so as not to impede normal powered plantar flexion movements. Finally, during the Swing state, a spring-damper PD control was implemented to provide toe clearance. The primary difficulty with the constant impedance control was the reduction of powered plantar flexion movements (**FIG. 8**). All data points for the normal participants are an average of 15 trials, whereas for the drop foot participants the average is over 20 trials. Here the spring-damper control used to prevent toe drag was acting against the foot when the users attempted to plantar flex their ankle during late stance.

[0059] The variable-impedance controller should have a similar maximum power plantarflexion angle as the zero impedance condition since both controllers were designed to not impede late stance power plantarflexion movements. However, this behavior was not observed (**FIG. 8**). It was discovered that the variable-impedance controller transitioned into the Swing state too early, before the foot actually left the ground, due to a lack of resolution in the forefoot force sensors. Consequently, the Swing spring-damper controller was activated too early, impeding power plantarflexion movements during late stance. In other embodiments, a foot switch can be positioned in the forefoot region to more accurately detect the event of toe-off.

[0060] In alternative embodiments, FES can be used to treat ankle foot gait pathologies, including drop foot gait. Instead of using a synthetic motor to vary ankle impedance, the muscles of the patient can be electrically stimulated to achieve desired ankle impedances as described herein. That is, a FES controller can be used to actively modulate ankle impedance to achieve a linear torsional spring during controlled plantar flexion to minimize forefoot collisions with